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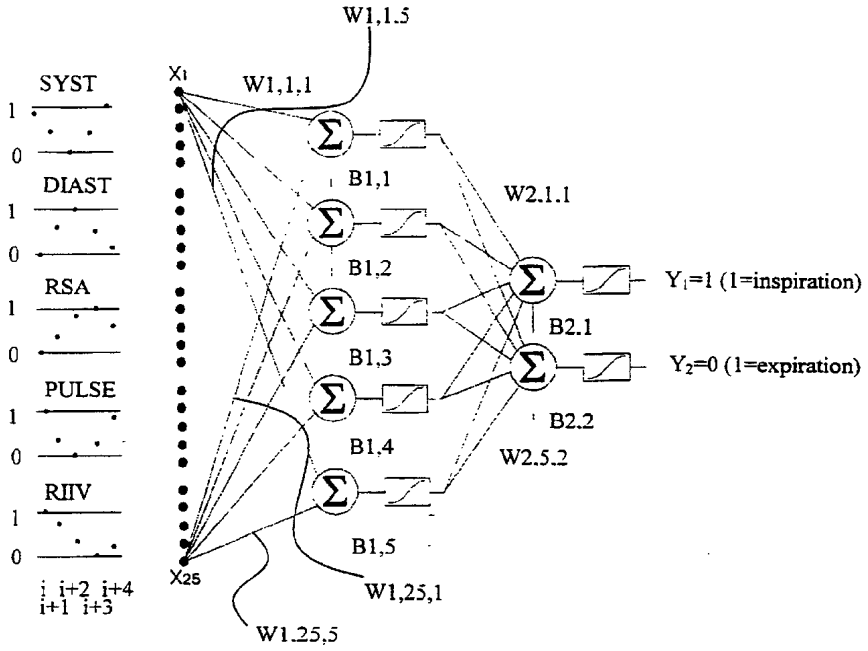
(54) Title: A METHOD FOR MEASURING INSPIRATION AND/OR EXPIRATION



(57) Abstract: The present invention relates to a method for measuring inspiration and/or expiration for photoplethysmographic (PPG) respiratory rate monitoring, comprising: detecting a PPG signal derivable from the inspiration and/or expiration from a patient by means of a detection device, extracting at least two different signal components from said PPG signal, wherein each of said different signal components is indicative of said inspiration and/or expiration, and by means of said extracted at least two different signal components deducing a respiratory synchronous signal.

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A METHOD FOR MEASURING INSPIRATION AND/OR EXPIRATION

FIELD OF INVENTION

The present invention generally relates to a method for measuring inspiration and/or expiration in photoplethysmographic (PPG) respiratory rate monitoring. Particularly, the present invention relates to utilisation of at least two signal components of a PPG signal for detection of inspiration and/or expiration.

BACKGROUND

Patients, who need to be observed, could among other things have their respiratory rate monitored.

One method of monitoring respiratory rate is to detect inspirations and expirations with photoplethysmographic (PPG). In SE 465551 it is described how to monitor the respiratory rate by using a PPG signal. However, this method is connected with a certain degree of erroneously detected inspirations and expirations, and therewith the respiratory rate is uncertain.

A photoplethysmographic (PPG) signal is obtained by measuring the absorption of light in blood and tissue. The dynamic part of the signal is caused by variations in the blood volume content of the tissue, but also, significantly, by blood perfusion [1]. The non-invasive character and the minimal disturbance of the variables under observation have made the technique widely used. The most widespread application of PPG is in monitoring of arterial oxygen saturation by pulse oximetry [2-3] wherein the cardiac synchronous component of the PPG signal is used. Other applications of the cardiac related component include cutaneous blood flow measurement and heart rate monitoring [1, 4-5].

A respiratory induced intensity variation (RIIV) of the PPG signal baseline is present [5]. The origin of the RIIV signal is not fully understood, but is believed to originate from the respiratory synchronous variations in intrathoracic pressure transmitted to the measurement site by the venous system [6-8]. Respiratory related components originating from the arterial side of the circulatory system are also present in the PPG signal. These include the well-known respiratory sinus arrhythmia (RSA) and the amplitude modulation of the cardiac pulse [9-11].

During respiration, the fluctuation of the intrathoracic pressure is affecting central venous pressure and peripheral venous pressure. This causes a blood volume displacement due to the high compliance of the venous system, which is seen in the PPG signal as a respiratory variation in its baseline (RIIV). As the venous return to the right ventricle increases during inspiration, the left ventricle output decreases due to interventricular dependence and blood pooling in the pulmonary circulation. After a delay of a few heartbeats, the blood reaches the left ventricle and causes the stroke volume to rise. This effect is seen as an amplitude modulation of the cardiac component of the PPG signal. These two effects are relatively invariable with time, but somewhat obscured by the RSA. In general, the heart rate is increasing after an inspiration. RSA is considered to reflect vagal control over heart rate, but its underlying mechanisms are not entirely clear.

In a neural network (NN), a structure of mathematical units (neurons) is constructed. Each neuron calculates the sum of its inputs and uses an activation function for its output. The neurons are connected by weights that are adjusted in a training process, aiming at minimizing the network error

function. During the training, the backward propagation algorithm is often used [14]. The main advantage of neural networks is the possibility to generate a complex decision boundary without knowing the statistical properties of the data under observation. This has made the method attractive in medical applications, and neural networks are used in fields such as diagnosing vascular disease from pulse waveforms [15] and ECG classification [16]. In ventilation monitoring, neural networks are used to detect breathing circuit problems [17], in ventilation mode recognition [18] and to classify effective or ineffective respiration in neonates [19].

SUMMARY OF THE INVENTION

An object of the present invention is to improve reliability in breath detection for respiratory rate monitoring.

This object among others is attained by a method as claimed in claim 1 and in claim 11, and a neural network as claimed in claim 9.

The respiratory synchronous signal deduced from at least two different signal components of a PPG signal exhibit an overall improved respiratory detection accuracy compared to a corresponding respiratory synchronous signal deduced from a single signal component, due to the different characteristics of the signal components.

Further characteristics of the invention and advantages thereof will be evident from the following detailed description of embodiments of the invention.

BRIEF DESCRIPTION OF THE DRAWINGS

The present invention will become more fully understood from the detailed description of embodiments of the present

invention given below and the accompanying figures, which are given by way of illustration only, and thus are not limitative of the present invention, wherein:

FIG. 1 shows a table of falsely detected respirations, detected with a zero-crossing algorithm,

FIG. 2 shows a table of falsely detected respirations, detected with a peak algorithm,

FIG. 3 shows a table of falsely detected respirations, detected with a neural network

FIG. 4 shows a measurement set-up,

FIG. 5 shows respiratory components of a PPG-signal,

FIG. 6 shows a neural network,

FIG. 7 shows an example of signal components,

FIG. 8 shows patterns for neural network analysis, and

FIG. 9 shows a neural network output.

DETAILED DESCRIPTION OF EMBODIMENTS

In the following description, for purposes of explanation and not limitation, specific details are set forth, such as particular techniques and applications in order to provide a thorough understanding of the present invention. However, it will be apparent to one skilled in the art that the present invention may be practiced in other embodiments that depart from these specific details. In other instances, detailed descriptions of well-known methods and apparatuses are omitted so as not to obscure the description of the present invention with unnecessary details.

A photoplethysmographic (PPG) signal includes respiratory

components seen as frequency modulation of the cardiac pulse (respiratory sinus arrhythmia, RSA), amplitude modulation (PULSE), systolic waveform (SYST), diastolic waveform (DIAST), and respiratory induced intensity variations (RIIV) in the PPG baseline.

The accuracy of these components (SYST, DIAST, RSA, PULSE, and RIIV) in respiratory rate measurements has been evaluated with the result according to the tables in Figs. 1-2, where the evaluation was performed with an arrangement as shown in Fig. 4. A PPG reflection mode sensor was positioned laterally on the forehead of a test object by using a non-elastic velcro headband, asserting negligible pressure on the sensor. The sensor included eight light emitting diodes emitting infrared light of 940 nm and two photo detectors. The components were mounted in a silicone bed and light barriers were included to avoid direct transmission of light trough the substrate. The detector-diode distances were 3.5 and 5 mm respectively. The detected signal was processed by offset balancing and amplification [5]. As a respiratory reference, airflow humidity was sensed by a fiber-optic sensor. This sensor detects respiratory variations in the humidity accumulating on an optical fiber end positioned in front of the airways [20]. The reference signal was delivered from the device as a square wave. The test object has the sensor in his nose to certify a correct detection of inspirations and expirations, and yet not impede respiration. Both the PPG and the reference signal were AD-converted and stored in the memory of a personal computer.

The respiratory induced intensity variations (RIIV) in the baseline of the PPG signal were extracted by using a 16^{th} order band-pass Bessel filter (0.13-0.48 Hz), and the cardiac related component by using a 5^{th} order band-pass Butterworth filter (0.50-2.0 Hz). Both filters were digital filters. From

the cardiac signal, each peak and valley were detected by using a simple algorithm based on the zero crossings of its derivative. For each cardiac pulse (denoted i), the following components were formed:

C_1: $SYST_i$ = Peak value (Systolic waveform)

C_2: $DIAST_i$ = Previous valley value (Diastolic waveform)

C_3: RSA_i = Time since last peak $^{-1}$ (Respiratory Sinus Arrhythmia)

C_4: $PULSE_i = SYST_i - DIAST_i$ (pulse amplitude)

Furthermore, a discrete version of the RIIV signal was signed as:

C_5: $RIIV_i$ = RIIV at time of $SYST_i$

Analyses of the five respiratory components C_j (j=1,…,5) were performed separately. The baseline trend of a 2-minute period was removed by subtracting the smoothed function of the signal, yielding $C'_{j,i}$ (where i is the heart beat number) according to:

$$C'_{j,i}=C_{j,i}-(C_{4j,i-4}+\ldots+C_{j,i+4})/9$$

Each period was also normalized to a mean value = 0 and standard deviation (SD) = 1 (yielding $C''_{j,i}$). Two simple algorithms were then applied for breath detection:

(I) Zero crossing algorithm inspiration (expiration) when:

$C''_{j,i}>0$ and $C''_{j,i-1}<0$

and the opposite for expiration (inspiration),

(II) Peak algorithm inspiration (expiration) when:

$C''_{j,i} > C''_{j,i-1}$ and $C''_{j,i} > C''_{j,i+1}$

and the opposite for expiration (inspiration).

Whether a true expression represents inspiration or expiration depends on the specific component. From the reference, each inspiration and expiration was noted, and the false positive (FP) and false negative (FN) breath detection rates were calculated for $j=1,\ldots,5$ and for both algorithms.

The forehead was chosen as measurement position as:

1) The photoplethysmograms from the forehead are accurately reflecting arterial and venous blood volume variations, as nervous effects are minimal [21]

2) In reflectance pulse oximetry, the forehead is an appropriate choice [22]

3) As the forehead is highly vascularised and the frontal bone acts as a reflector, photoplethysmograms of high amplitude and signal-to-noise ratio are obtained [23-24], and

4) The sensitivity to patient motion is reduced at this position.

For these reasons, a traditional transmission measurement approach was not considered. Measurements were performed during spontaneous and metronome controlled breathing. The latter was used to produce different respiratory rates and breath-to-breath intervals, and thus, a more variable and realistic material.

When attempting to combine the respiratory components, pattern recognition in the form of a neural network was chosen [14], which has two advantages:

1) As the relation between the different respiratory

components of the PPG signal is complex and their statistical properties are not fully known, the decision boundary can be formed by the network, and

2) The implementation in a monitoring system is straightforward and the speed of processing (once trained) is high.

In order to structure the network, empirical rules and experience is required. The inputs to the present network consist of five well-known components of the PPG signal at five consecutive heartbeats after each respiratory event (inspiration/expiration or neither of the two). Five heartbeats represent a reasonable amount of time for the effects after each event to be fully captured. One hidden layer of neurons was assumed sufficient for this problem. This layer used 5 neurons, chosen from the rule of thumb saying that the number of hidden neurons should be the 2-logarithm of the number of inputs [19]. Standard error function minimization was performed by backward propagation. The training was stopped after a fixed number of epochs, and restarted if the error was above a specific level. This procedure was used to avoid local minima and to not over-fit the network to the training set. By shifting the signals one heartbeat a new pattern was formed. This caused each value to be seen by the network five times. However, as a new pattern was generated after each shift, effects of redundancies are supposed to have a minor influence on the results.

The PULSE parameter is actually a linear combination of two other inputs. This parameter was included anyhow, as it was plausible that this form of pre-processing would make it easier for the network to converge. This was confirmed when this parameter made the least number of errors in the separate analysis.

The patterns following expiration were less clear than those following inspiration. Therefore, only the first output of the network, representing inspiration, was used in the network evaluation. It is possible to further improve the system by combining the two outputs. Another important part was the determination of a suitable threshold to interpret the network output automatically. In the present evaluation, this threshold was set manually.

In the table in Fig. 1 the first column shows the number of errors, for false positives FP, for all test subjects as a percentage of the total number of reference breaths. The second column shows the error rate (SD), for false positives FP, as calculated on individual basis. The third and fourth columns show corresponding percentages for false negatives FN. The calculations in the table in Fig. 1 were analyzed with the Zero crossing algorithm.

The table in Fig. 2 shows corresponding values to those shown in Fig. 1 but in this case the calculations were analyzed with the Peak algorithm.

The best result, i.e. the lowest overall error rate, was found for the PULSE component analyzed by the Zero crossing algorithm (the PULSE row). In a preferred combination of PPG signal components, the PULSE component is one of the components.

Fig. 3 shows a table where the result is evaluated with a neural network. The number of false positive FP and false negative FN breath detections are presented for both natural and metronome measurements together with the total number of reference breaths n and the threshold tr used in the detection process.

In Fig. 5 it is shown an example of a PPG signal and how the

different signal components are defined. Extraction of the respiratory components of the PPG signal with five values from each cardiac pulse (denoted i): the systolic value $SYST_i$, the diastolic value $DIAST_i$, the respiratory sinus arrhythmia RSA_i, the pulse amplitude $PULSE_i$, and the RIIV signal $RIIV_i$.

The general effects of an inspiration seem to be a decrease in SYST and an increase in DIAST. This causes the PULSE parameter to fall. RSA is increasing after an inspiration and RIIV is decreasing. The opposites are seen at expiration.

A fully connected feed-forward neuron network is shown in Fig. 6, in which all five signal-components are used as inputs. Each signal component has five input neurons, the hidden layer has five neurons, and there are two output neurons Y1, Y2. The top output neuron Y1 indicates, with a 1, if inspiration occurs, and the lower output neuron Y2 indicates, with a 1, if expiration occurs. At a specific time (i) and event (inspiration, expiration, or neither of the two) five consequent values of each component were normalized and used as inputs.

The components C_1-C_5 were presented to the neural network as the values at five consecutive heartbeats at a specific time (i). The five values of each component were normalized to a range of 0-1. An output of [1 0] was assigned when the inputs followed inspiration, [0 1] when the inputs followed expiration and [0 0] when following neither of the two. Shifting the five signals one heartbeat, performing a new normalization and assigning a new output formed the next pattern. In this way, typically 60 patterns per minute measurement time were obtained (or more accurately, the number of heartbeats during the minute minus four). Consequently, a network with twenty-five inputs and two outputs was obtained. The network was a fully connected feed-forward network, with

sigmoid activation functions and bias values used throughout.

In situations where some components are weak or disturbed (e.g. a less pronounced RSA in an elderly patient) the network will recognize the respiration from the other components. To do this effectively, it is important that the neural network has been trained by relevant data recorded in the specific environment (less pronounced RSA in the exemplary case).

Fig. 7 shows characteristic one-minute example (in metronome controlled breathing) of a cardiac signal, a RIIV "raw signal", and a reference signal (time as x-axis). Inspirations are marked as dots in the diagram. The five calculated respiratory components C_1-C_5 are included (heartbeat number as x-axis) along the y-axis in arbitrary units.

The patterns for the neural network analysis are shown in Fig. 8. The boxes show for each component lower quartile, median and upper quartile values. The patterns are divided according to the output (inspiration, expiration, and others). The x-axes are heartbeat number after the event.

A typical two-minute neural network output (solid line) is shown in Fig. 9, i.e. a respiratory synchronous signal, where peaks above the threshold (dashed line) are regarded as inspirations. Reference inspiration points are marked in the figure, and the x-axis shows the heartbeat number.

Advantage of applying the PULSE component is that it originates from the "high-pressure" arterial system and is therefore less affected by environmental and physiological disturbances. In comparison, the RIIV signal, originating from the "low-pressure" venous system, is less accurate. However, it is still included as it is largely different in origin to the other components, which is a positive aspect in the pattern recognition. Further developments, focused on the most

accurate components will optimize a network structure.

To achieve a reliable respiratory rate the different PPG signal components are utilized in dependence on how well each respective component contributes to an accurate detection of inspirations or expirations. Detection of respirations is resolved in time in several parts, where each different component part is weighed against its probability of accurate detection of inspiration or expiration. The PULSE component, as identified above as the most accurate component, will probably provide the most often used component part. As complement, the other components will be used, when better identifying inspirations or expirations. A possible way of achieving this is to utilize a neural network as above described.

An advantage with the estimation of a respiratory rate by means of PPG signal components is the possibility to combine this with the estimation of arterial oxygen saturation (pulse oximetry) and/or cardiac pulse frequency by means of PPG.

It is also to be noted that the amplitude of the respiratory rate signal components may be used in order to estimate tidal volumes.

It has above been described that the respiratory rate has been estimated by means of optical signals. It is also possible to utilize the same signal components by means of other physiological signals, e.g. skin impedance measurements or blood pressure measurements.

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CLAIMS

1. A method for measuring inspiration and/or expiration in photoplethysmographic (PPG) respiratory rate monitoring, comprising:

- detecting a PPG signal derivable from the inspiration and/or expiration from a patient by means of a detection device,

characterised by

- extracting at least two different signal components from said PPG signal, wherein each of said different signal components is indicative of said inspiration and/or expiration, and

- by means of said extracted at least two different signal components deducing a respiratory synchronous signal.

2. The method according to claim 1, comprising using each of said at least two signal components in dependence on their respective accuracy in deducing of said signal indicative of said inspiration and/or expiration.

3. The method according to claim 2, further comprising:

- dividing each of the at least two signal components into a number of segments, and

- deducing said signal indicative of said inspiration and/or expiration, by means of, for each segment, choosing the signal component that provides the best accuracy.

4. The method according to any preceding claim, comprising using five different signal components detecting inspiration and/or expiration.

5. The method according to any preceding claim, wherein each signal component is anyone of respiratory sinus arrhythmia (RSA), respiratory induced intensity variations (RIIV), systolic waveform (SYST), diastolic waveform (DIAST), or pulse amplitude (PULSE).

6. The method according to any preceding claim, wherein the detection device is applied to the skin of the patient.

7. The method according to any preceding claim, wherein the detection of the PPG signal is performed during five consecutive heartbeats, for each deduction of inspiration and/or expiration.

8. A method for photoplethysmographic (PPG) respiratory rate monitoring c h a r a c t e r i s e d b y

- measuring inspiration and/or expiration according to a method as claimed in any preceding claims during a predetermined period of time,

- counting inspiration and/or expiration from said deduced signal indicative of inspiration and/or expiration,

- deducing a respiratory rate by means of said counted number of inspiration and/or expiration, and

- indicating the thus obtained respiratory rate.

9. A neural network implementing the method according to any of claims 1 to 7.

10. A method of training the neural network as claimed in claim 9, comprising a plurality of training processes, and each training process is ended in dependence of a percentage of erroneous detections exceeding a predetermined value.

11. A method for measuring inspiration and/or expiration in respiratory rate monitoring, comprising:

- detecting a signal derivable from the inspiration and/or expiration from a patient by means of a detection device,

characterised by

- extracting at least two different signal components from said signal, wherein each of said different signal components is indicative of said inspiration and/or expiration, and

- by means of said extracted at least two different signal components deducing a respiratory synchronous signal.

	FALSE POSITIVES		FALSE NEGATIVES	
SYST	11.0%	11.8%(0.7)	3.5%	3.2%(0.3)
DIAST	7.5%	8.5%(0.7)	3.3%	3.0%(0.4)
RSA	3.2%	3.7%(0.4)	7.7%	6.9%(0.5)
PULSE	4.5%	5.2%(0.5)	5.2%	4.7%(0.4)
RIIV	6.8%	7.7%(0.5)	6.6%	5.9%(0.4)

FIG. 1

	FALSE POSITIVES		FALSE NEGATIVES	
SYST	27.8%	30.3%(1.7)	0.4%	0.4%(0.04)
DIAST	20.1%	22.9%(1.9)	0.8%	0.8%(0.1)
RSA	11.2%	12.5%(1.1)	2.2%	2.1%(0.2)
PULSE	12.9%	14.7%(1.3)	1.1%	1.1%(0.1)
RIIV	17.0%	18.5%(1.0)	2.1%	1.9%(0.1)

FIG. 2

SUBJECT	NATURAL BREATHING				METRONOME BREATHING		
	n	FP	FN	tr	n	FP	FN
1	93	1	1	0,05	84	1	3
2	71	0	0	0,1	65	4	1
3	55	3	0	0,1	64	0	0
4	72	4	4	0,1	70	6	9
5	46	4	0	0,15	69	1	1
6	57	3	0	0,18	66	2	0
7	62	3	9	0,4	67	6	2
8	106	1	1	0,05	69	5	4
9	71	5	6	0,5	63	8	6
10	76	1	0	0,1	73	1	1
11	58	5	0	0,1	67	4	1
12	97	1	7	0,1	77	4	0
13	64	5	0	0,4	73	2	4
14	29	10	5	0,4	69	2	6
15	36	16	1	0,5	72	0	15
TOTAL		62	34		TOTAL	46	53
		6.2%	3.4%			4.4%	5.1%
			n=993				n=1048

FIG. 3

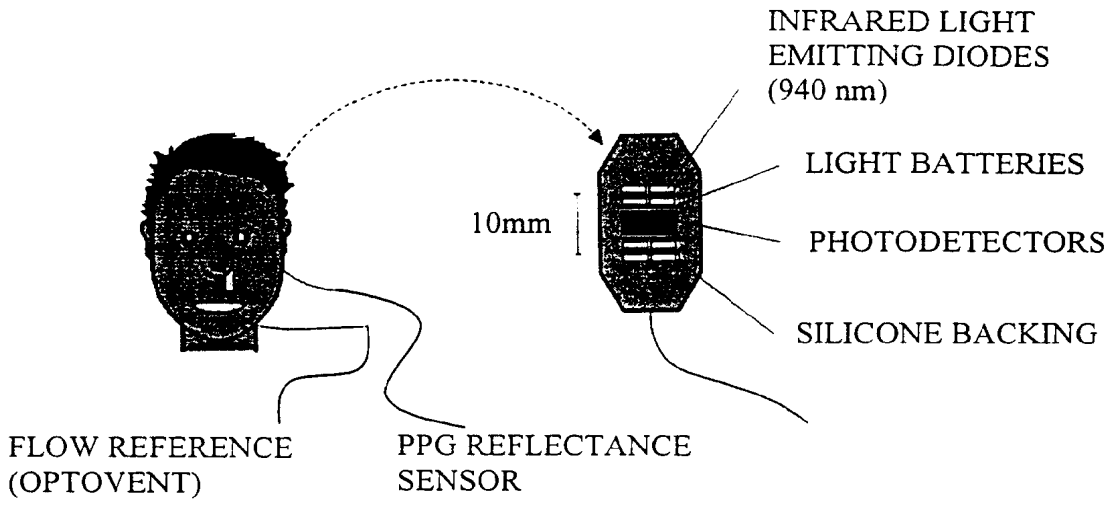
INFRARED LIGHT
EMITTING DIODES
(940 nm)
LIGHT BATTERIES
10mm
PHOTODETECTORS
SILICONE BACKING
FLOW REFERENCE
(OPTOVENT)
PPG REFLECTANCE
SENSOR

FIG. 4

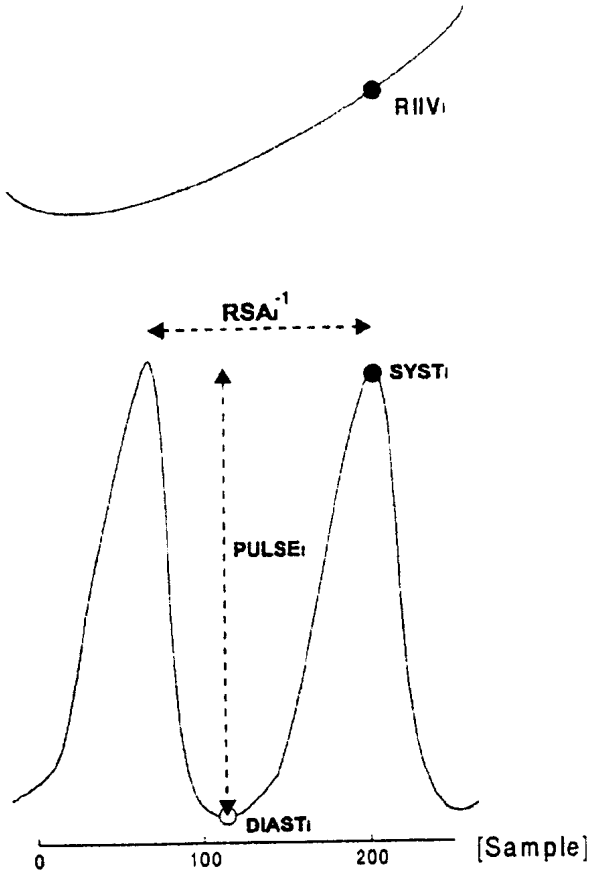
RIIVi
RIIV signal
RSAi-1
SYSTi
PULSEi
DIASTi
0
100
200
[Sample]
Cardiac signal

FIG. 5

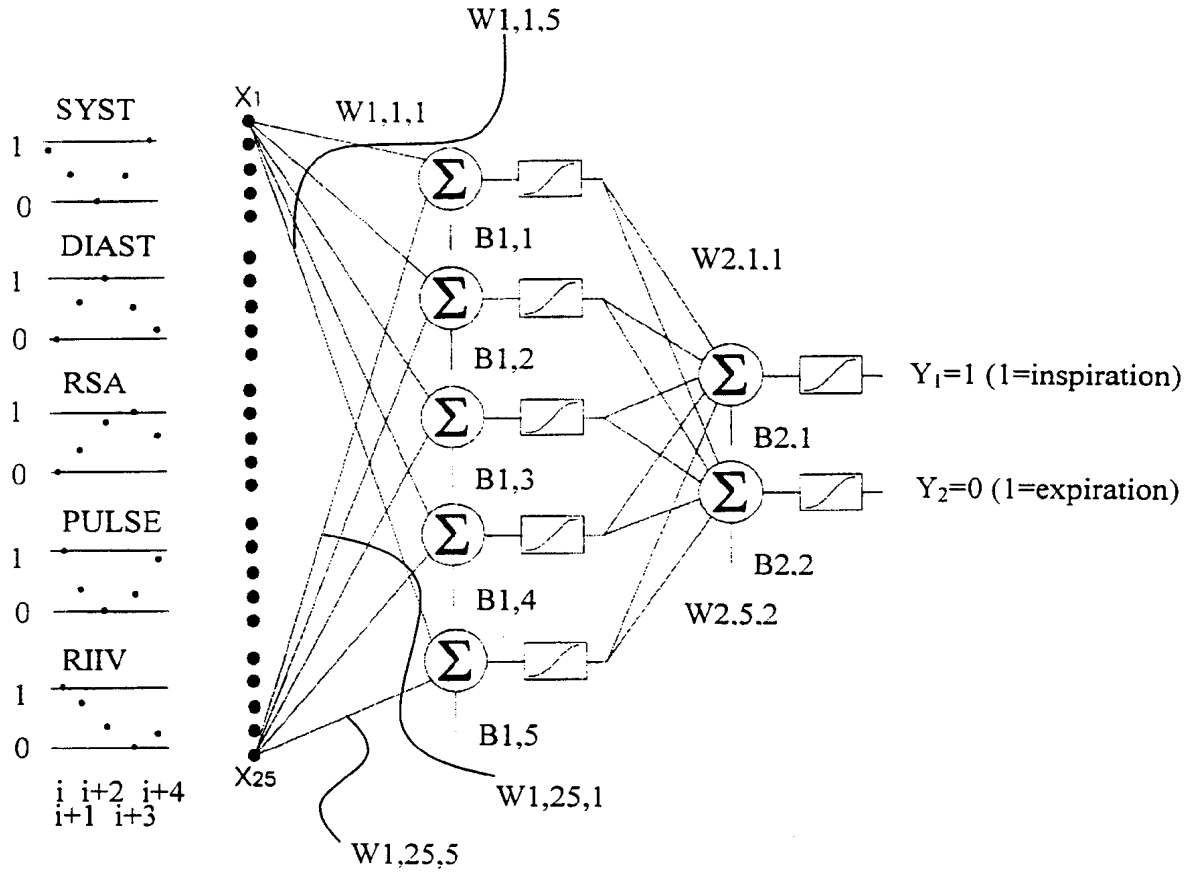
W1,1,5
SYST
1
0
DIAST
1
0
RSA
1
0
PULSE
1
0
RIIV
1
0
i i+2 i+4
i+1 i+3
X1
W1,1,1
B1,1
B1,2
B1,3
B1,4
B1,5
W2.1.1
B2.1
B2.2
W2.5.2
Y1=1 (1=inspiration)
Y2=0 (1=expiration)
X25
W1,25,1
W1,25,5

FIG. 6

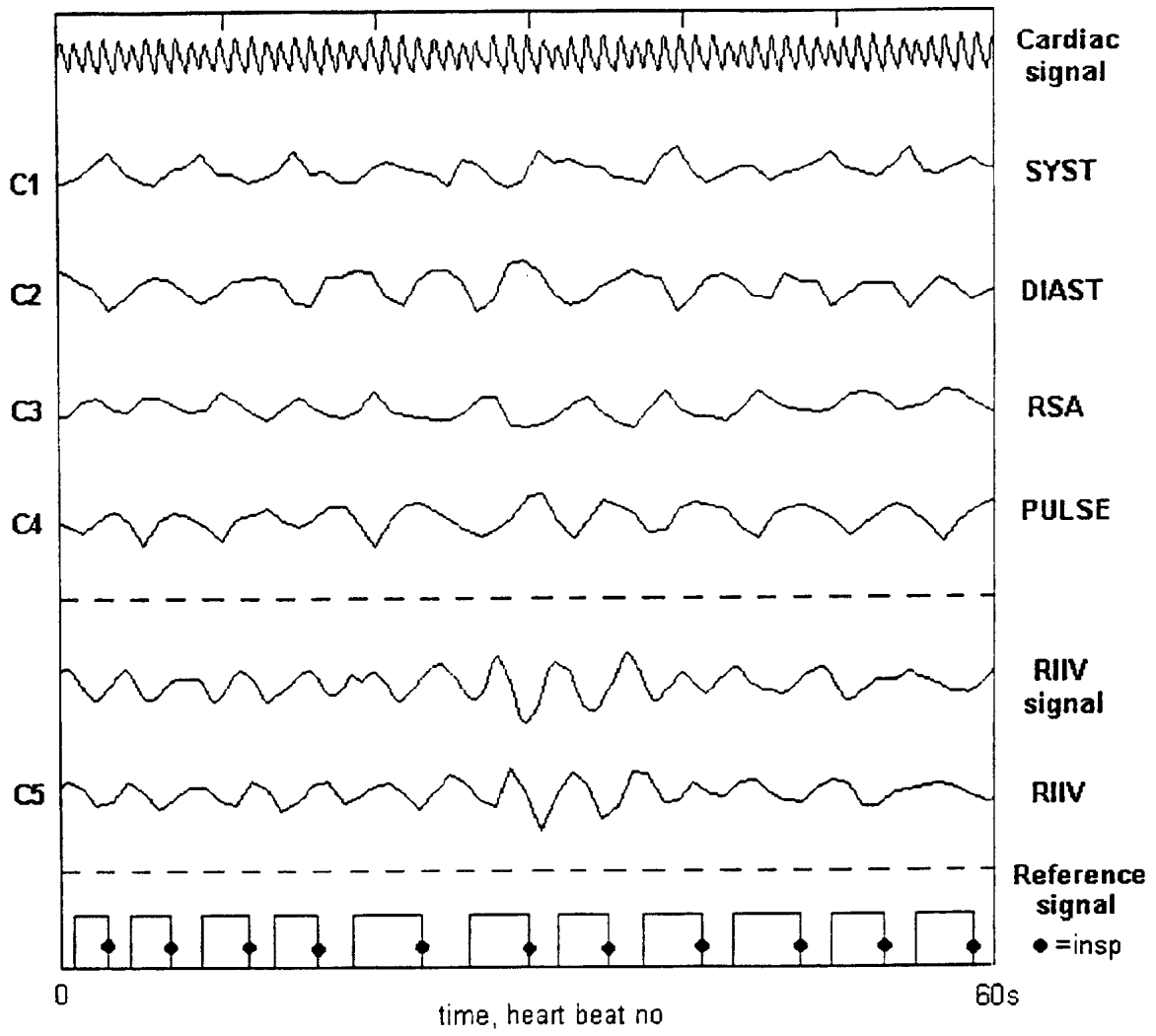
Cardiac signal
C1
SYST
C2
DIAST
C3
RSA
C4
PULSE
RIIV signal
C5
RIIV
Reference signal
● =insp
0
60s
time, heart beat no

FIG. 7

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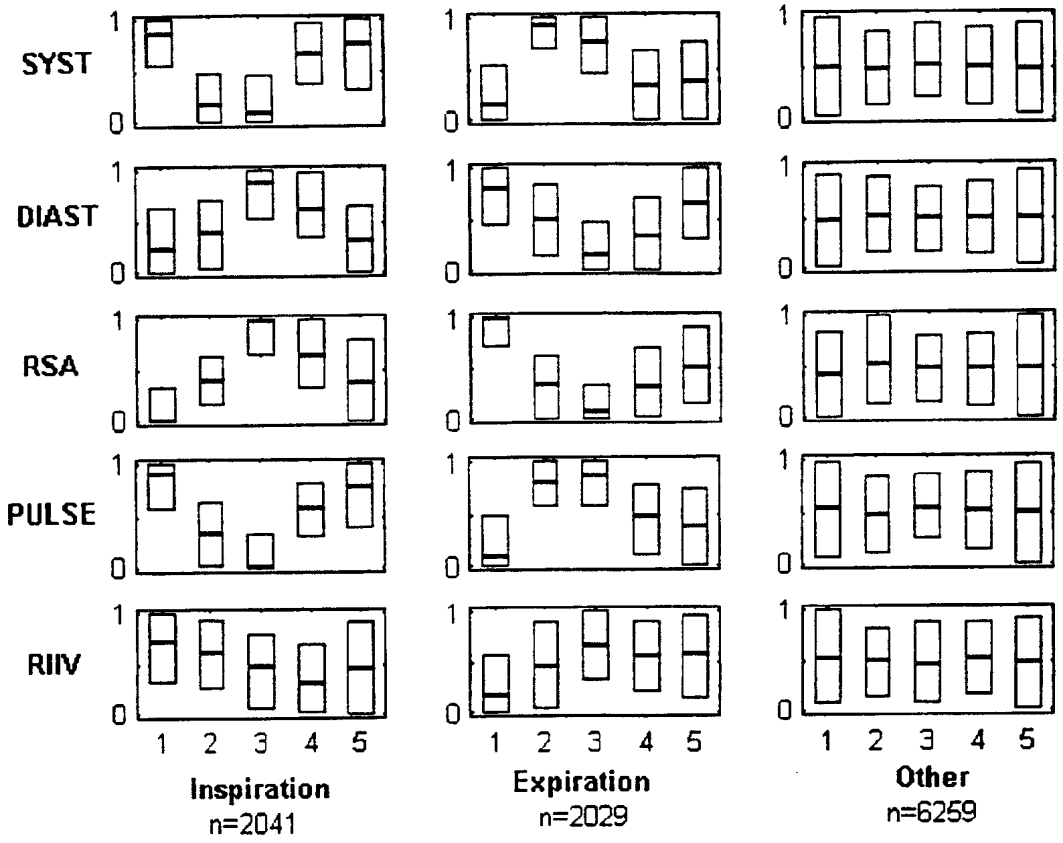
SYST
DIAST
RSA
PULSE
RIIV
1
0
1 2 3 4 5
Inspiration
n=2041
Expiration
n=2029
Other
n=6259

FIG. 8

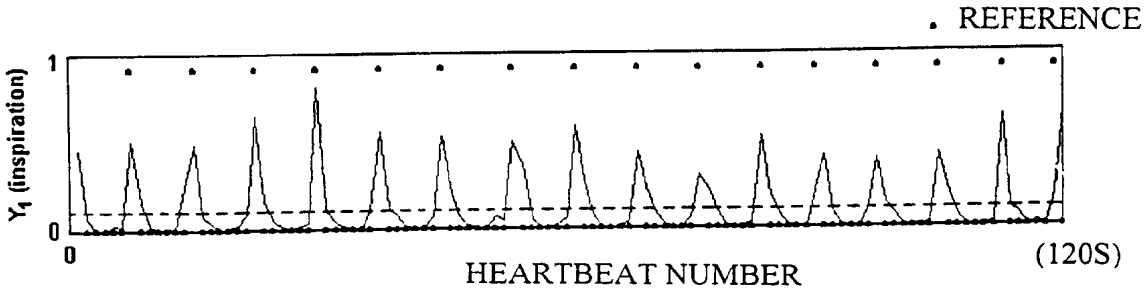
. REFERENCE
Y_1 (inspiration)
1
0
0
(120S)
HEARTBEAT NUMBER

FIG. 9

INTERNATIONAL SEARCH REPORT

International application No.
PCT/SE 01/00782

A. CLASSIFICATION OF SUBJECT MATTER

IPC7: A61B 5/0205
According to International Patent Classification (IPC) or to both national classification and IPC

B. FIELDS SEARCHED

Minimum documentation searched (classification system followed by classification symbols)

IPC7: A61B, A61M

Documentation searched other than minimum documentation to the extent that such documents are included in the fields searched

SE,DK,FI,NO classes as above

Electronic data base consulted during the international search (name of data base and, where practicable, search terms used)

C. DOCUMENTS CONSIDERED TO BE RELEVANT

Category*	Citation of document, with indication, where appropriate, of the relevant passages	Relevant to claim No.
X	SE 465551 B (LARS-GÖRAN LINDBERG ET AL.), 30 Sept 1991 (30.09.91), see page 5-7,17,24	1-2,6-7,9-11
A		3-5,8
	--	
A	WO 0015106 A1 (JAY, GREGORY, D.), 23 March 2000 (23.03.00), page 4, line 24 - page 5, line 13	1-11
	--	
A	US 5273036 A (HARALD KRONBERG ET AL.), 28 December 1993 (28.12.93), column 3, line 51 - column 4, line 57, abstract	1-11
	--	

☐ Further documents are listed in the continuation of Box C. ☒ See patent family annex.

* Special categories of cited documents:
"A" document defining the general state of the art which is not considered to be of particular relevance
"E" earlier application or patent but published on or after the international filing date
"L" document which may throw doubts on priority claim(s) or which is cited to establish the publication date of another citation or other special reason (as specified)
"O" document referring to an oral disclosure, use, exhibition or other means
"P" document published prior to the international filing date but later than the priority date claimed
"T" later document published after the international filing date or priority date and not in conflict with the application but cited to understand the principle or theory underlying the invention
"X" document of particular relevance: the claimed invention cannot be considered novel or cannot be considered to involve an inventive step when the document is taken alone
"Y" document of particular relevance: the claimed invention cannot be considered to involve an inventive step when the document is combined with one or more other such documents, such combination being obvious to a person skilled in the art
"&" document member of the same patent family

Date of the actual completion of the international search	Date of mailing of the international search report
3 Sept 2001	05 -09- 2001
Name and mailing address of the ISA/ Swedish Patent Office Box 5055, S-102 42 STOCKHOLM Facsimile No. +46 8 666 02 86	Authorized officer Ulrika Westman/AE Telephone No. +46 8 782 25 00

INTERNATIONAL SEARCH REPORT

Information on patent family members

02/08/01

International application No.
PCT/SE 01/00782

Patent document cited in search report	Publication date	Patent family member(s)	Publication date
SE 465551 B	30/09/91	CA 2075064 A	17/08/91
		DE 69113785 D,T	02/05/96
		EP 0515482 A,B	02/12/92
		JP 5504084 T	01/07/93
		SE 9000564 A	17/08/91
		US 5396893 A	14/03/95
		WO 9111956 A	22/08/91
WO 0015106 A1	23/03/00	AU 6034799 A	03/04/00
		EP 1112023 A	04/07/01
		US 6129675 A	10/10/00
US 5273036 A	28/12/93	NONE	